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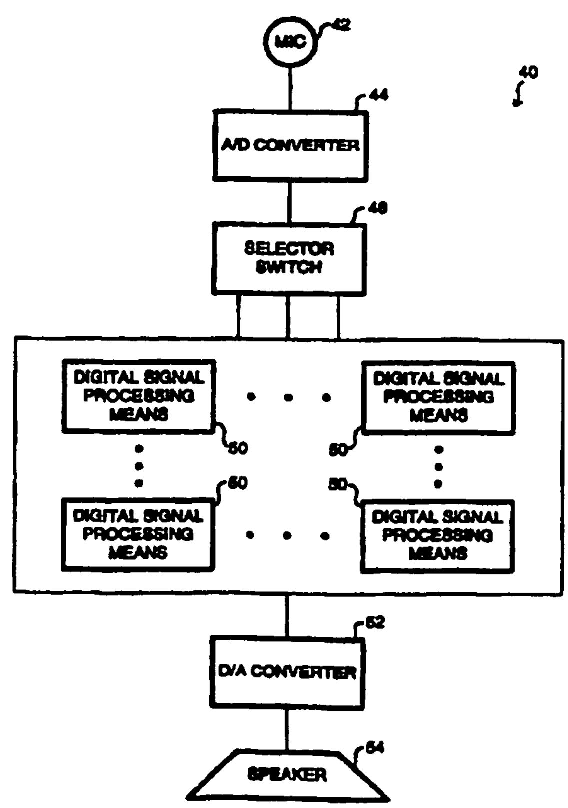
(54) Title: DIGITAL SIGNAL PROCESSING HEARING AID WITH PROCESSING STRATEGY SELECTION

(57) Abstract

A digital signal processing hearing aid is disclosed having a plurality of digital signal processing means (50) for processing input digital signals, and a selector switch (46) manipulable by a user for choosing which of the processing means to utilize. Each of the digital signal processing means is designed to provide optimal results in a particular listening environment. Since the user is allowed to choose which of the plurality of processing means to invoke, and since each processing means is specifically designed to operate in a particular listening environment, the hearing aid is capable of providing excellent results in a plurality of listening environments.



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DIGITAL SIGNAL PROCESSING HEARING AID WITH PROCESSING STRATEGY SELECTION

Background of the Invention

A common problem associated with sensorineural hearing loss is recruitment. A hearing impaired person suffering from recruitment has an elevated threshold for soft sounds. This means that soft sounds which are audible to a person with normal hearing will have to be made louder in order to be heard by the hearing impaired person. However, with recruitment, loud sounds may be just as loud for the hearing impaired person as for the person with normal hearing. This represents a loss of dynamic range for the hearing impaired. This loss of dynamic range may vary with frequency. For example, at low frequencies the hearing impaired person may have nearly the same dynamic range as the person with normal hearing, but at high frequencies the dynamic range of the hearing impaired person may be considerably reduced. This impaired dynamic range is often referred to as the residual dynamic range.

The loss of dynamic range in the hearing impaired is most often attributed to malfunction of the outer hair cells of the cochlea. Sound vibrations in the air are transmitted from the ear drum and through the ossicles of the middle ear to the inner ear and the cochlea. Inside the cochlea are the flexible tectorial membrane and the more rigid basilar membrane. Between these two membranes lie the inner and outer hair cells. Ninety-five percent of the afferent neural fibers which transmit acoustic information to the brain are connected to the inner hair cells. The longest cilia of the outer hair cells are connected to the tectorial membrane, but the inner hair cells have no such connection. Both the inner and outer hair cells are connected to the basilar membrane through supporting cells. Vibrations passing between the tectorial and basilar membranes cause more motion in the flexible tectorial membrane than in the basilar membrane. This difference in motion causes a sheering motion along the outer hair cells. The outer hair cells react to this shearing motion in a complex manner. The entire mechanism is not yet clearly understood but it appears that the outer hair cells stretch and contract according to the intensity of the vibrations in a manner which amplifies these vibrations. For larger amplitude vibrations, however, the outer hair cell motion saturates causing a reduction in amplification. This nonlinear, saturating amplification corresponds to a natural dynamic range compression. The compressed vibrations from the outer hair cells are communicated to the inner hair cells and then through the afferant neural fibers to the brain. When the outer hair cells malfunction, there is a loss of natural compression and recruitment occurs. The inner hair cells may continue to functions normally and there may be a mild to moderate hearing loss. More severe hearing losses will occur with loss of inner hair cell function.

Many hearing aid instruments have been designed to deal with this problem. The approach taken is to compress the dynamic range of the input sound signal so that it more nearly fits into the residual dynamic range of the recruited ear. The ratio of input dynamic range in dB to compressor output dynamic range in dB is called the compression ratio. To adequately specify

the compressor, the compression ratio needs to be accompanied by a static gain value. This static gain value will determine at which input power level the system delivers a specified fixed gain. For example the static gain may be set so that at 80dB SPL input power, the system delivers unity gain. If the compressor is set to a 2:1 compression ratio, then at 60dB SPL input power the system will produce a 70dB SPL output, that is a gain of 10dB, and at 100dB SPL input power the system will produce a 90dB SPL output, that is a gain of -10dB.

Usually the compression ratio is not constant over the entire input power range. A low level compression knee may be defined. For input powers below this low level compression knee, the compression ratio may be 1:1, that is, a fixed linear gain may be applied. The designated compression ratio (e.g. 2:1) may take affect only for input power levels above this low level compression knee. A high level or limiting knee may also be defined. For input power levels above this high level knee, the compression ratio may increase or even become infinite, or it may be that the output level is fixed regardless of increase in input level. A system which has only a high level compression knee below which the compression ratio is 1:1 (linear gain) is called a limiter. A system which has a low level compression knee positioned at 40-50dB SPL is termed a full range compressor.

Even without reference to the electro-mechanics of the inner ear and the natural loss of compression due to malfunction of the outer hair cells, the need for compressors or limiters in hearing aids has long been recognized. The need for hearing aids to have large gains to make softer sounds audible has driven amplifiers and output transducers out of their linear ranges. Earlier hearing aids accomplished limiting by letting the amplifier and/or output transducer clip. Unfortunately this caused harmonic distortion which, at high frequencies, masked softer speech sounds and generally reduced fidelity in the system (See M. C. Killian. *The K-Amp Hearing Aid: An Attempt to Present High Fidelity for Persons with Impaired Hearing*, American Speech-Language-Hearing Association, July 1993, at 52-74). Later systems introduced limiters to help alleviate this problem, and still later systems used full range dynamic range compression (See e.g. Fred Waldhauer et al., *Full Dynamic Range Multiband Compression in a Hearing Aid*, The Hearing Journal, Sept. 1988, at 1-4).

The compression process requires a means for measuring the power of the input signal and generating a dynamically varying gain as a function of this input power. This gain is then applied to the signal which is delivered to the ear. When the input power is low, this gain will generally be high so that soft sounds are made louder. When the input power is high, this gain will generally be low so that loud sounds are not made too loud. The measure of input power requires averaging over time. The time span of the averaging defines a compression time constant. If the time span is very long then the compressor will react slowly to changes in input power level. This is sometimes referred to as Automatic Gain Control (AGC) where time constants of one to two seconds are typical. When the time span of the averaging is short the compressor will react quickly to changes in input power level. With a time span of approximately five to fifty milliseconds, the compressor may be referred to as a syllabic rate

compressor. A syllabic rate compressor will limit the gain of a loud vowel sound while amplifying a soft consonant which immediately follows it.

In most designs there is both an attack and release compressor time constant. The attack time constant determines the time it takes for the compressor to react at the onset of a loud sound. That is, the time it takes to turn down the gain. The release time constant determines the time it takes for the system to turn up the gain again after the loud sound has terminated. Most often the attack time is quite short (<5 milliseconds) with the release time being longer (anywhere from 15 to 100s of milliseconds).

Even with separate attack and release times, there have still been problems with compressor time constants. With a long release time, any short impulse in the room (e.g. the clank of a dish) will cause the gain to be shut down for the length of the relatively long release time. On the other hand, if the time constant is always short, it will cause an annoying swell in volume every time a speaker takes a breath. This problem has been alleviated by the introduction of adaptive time constants. Hotvet introduced in U.S. Patent No. 4,718,499 an adaptive time constant system in which the release time constant for a loud sound in silence is short but the release time constant gradually becomes longer proportional to the length of the louder sounds in the environment. Thus, if a speaker speaks in a normal rhythm, the release time constant will grow longer, reducing the amplitude swell in the brief silences between words. Others have also discussed multiple time constant systems with a similar goal in mind (See e.g. R. F. Laurence, et al., *A Comparison of Behind-the Ear High-Fidelity Linear Hearing Aids and Two-Channel Compression Aids, in the Laboratory and in Everyday Life*, Br. J. Audiol., 1983, at 17:31-48; and Brian Moore, et al., *Optimization of a Slow-Acting Automatic Gain Control System for Use in Hearing Aids*, Br. J. Audiol., 1991, at 25:171-182).

To match the variability of recruitment with frequency, a compressor is often designed to perform differently in different frequency bands. A multi-band compressor divides the input signal into multiple frequency bands and then measures power in each band and compresses each band separately with possibly different compression ratios and time constants in the different bands. For example a properly designed two band compressor can make soft high frequency consonants audible while suppressing low frequency competing noises occurring simultaneously. Vilchur (See E. Vilchur, *Signal Processing to Improve Speech Intelligibility in Perceptive Deafness*, J. Acoust. Soc. Am. 53, 1973, at 1646-1657) discussed a bench top prototype of a two band compressor. Barfod (See J. Barfod, *Multichannel Compression Hearing Aids*, Report No. 11, The Acoustic Laboratory, Technical University of Denmark, 1976) discussed compressors of up to four bands. These compressors also had variable time constants in the different frequency bands.

The outer hair cells of the cochlea, when functioning normally, are often thought to perform compression function in overlapping frequency bands called critical bands. These frequency bands are spaced linearly at intervals of approximately 100 Hz at frequencies below about 500 Hz, and are spaced logarithmically at approximately third octave intervals above

500Hz. Thus, the outer hair cells behave as a biological critical band compressor. The time constant associated with this compressor has been approximated to be about 1ms. Lippman et. al. (See R. P. Lippman, et al., *Study of Multichannel Amplitude Compression and Linear Amplification for Persons with Sensorineural Hearing Loss*, J. Acoust. Soc. Am. 69(2), Feb. 1981, at 524-534) designed a benchtop 16 band compressing hearing aid system with the bands tuned to match the critical bands of hearing. Each band represented a separate compression channel. Two settings of this compressor were compared against a linear non-compressing system. Martin (See G. R. Martin, *Studies of Real-Time Multiband Adaptive Gain Hearing Aids*, MIT, Sept. 1992, at 1-103) discussed a 3rd octave band compression hearing aid system using digital signal processing.

As the number of compression bands increases, each with its own compression ratio and static gain, it is possible to view the compressor as having an almost continuously varying compression ratio as a function of frequency. In this case the system may be represented as a set of frequency dependent gain curves. Each gain curve applies at a certain input power level. For input between these power levels, the system interpolates between gain curves. Killian (previously cited) discusses the K-amp hearing aid system which integrates power in one band but uses the power estimate to interpolate between low level and high level frequency response curves. The low power level frequency response curve has generally more gain and, in particular, more gain at high frequencies then at low. The high power level frequency response curve has generally less gain and is more flat across frequencies. There is an optional setting which allows the low power level curve to also be set flat.

The process of adjusting the compression ratios or gain curves of a compressor is central to the hearing aid fitting process. One approach to doing this is to attempt to adjust the compressor so that for all input levels and all frequencies the hearing impaired listener has the same impression of loudness that a normal listener would have. Loudness is a perceptual quantity which can under certain constraints be plotted as a function of input power level. The loudness growth curve may be measured by presenting a number of input signals at different levels and asking the listener to subjectively rate these on a perceptual scale (e.g. 1 to 10). By measuring the loudness growth curves of an impaired listener at different frequencies and comparing these to the loudness growth curves of an average of normal listeners, a loudness matching compression fitting can be attempted. To accurately match loudness growth curves, the hearing instrument would permit continuously variable compression ratio over input level. In this case it is more useful to think in terms of continuously variable input/output power curves. The system described above with low and high level compression knees is able to implement only three segment piecewise input output curves. Barfod (previously cited) and Lippman et. al. (previously cited) attempted to fit their multi-band compression systems so as to restore the loudness growth curves of the impaired ear to match those of the normal ear.

Loudness matching compression fitting has its limits. If the recruited ear has 5dB of residual dynamic range it will not be effective to compress a 90 dB input dynamic range into this

5dB. Instead, some amount of compression will be applied and then a static gain defined so that the most useful part of the input dynamic range (e.g. typical speech range) is roughly centered in the residual dynamic range. Limiting will be applied for louder signals. Finding good compromises in fitting compressors is central to the art of hearing aid fitting.

There has been some discussion about whether it is indeed necessary to test the loudness growth curves of the impaired listener as part of the fitting process or whether it is possible to predict them from the threshold audiograms. Kollmeier et al. (See B. Kollmeier, et al., *Speech Enhancement by Filtering in the Loudness Domain*, Acta Otolaryngol (Stockh) 1990, Suppl. 469:207-214) has shown that the shape of loudness growth curves becomes less predictable with increasing hearing loss. That is, the variance between subjects increases with hearing loss. This indicates that successful prediction from the threshold is unlikely.

There has been much discussion regarding the nature of improvements due to compression. Vilchur (previously cited) and Yanick (See P. Yanick, *Effects of Signal Processing on the Intelligibility of Speech in Noise for Subjects Possessing Sensorineural Hearing Loss*, J. Am. Audiol. Soc. 1, 1976, at 229-238) showed improvements in intelligibility with their compression systems, while Abramovits (See R. Abramovits, *The Effects of Multichannel Compression Amplification and Frequency Shaping on Speech Intelligibility for Hearing Impaired Subjects*, Unpublished doctoral thesis, City University of New York, 1979), Mangold et al. (See S. Mangold, et al., *Programmable Hearing Aid with Multichannel Compression*, Scand. Audiol. 8, 1979, at 121-126), O'Loughlin (See B. O'Loughlin, *Evaluation of a Three Channel Compression Amplification System on Hearing-Impaired children*, Aust. J. Audiol. 2, 1980, at 1-9), and Lippman et al. (previously cited) failed to show intelligibility improvements. It has also been argued in Moore (See Brian Moore, *Evaluation of a Dual-Channel Full Dynamic Range Compression System for People with Sensorineural Hearing Loss*, Ear and Hearing, Vol. 13, No. 5, 1992, at 349-370) that it is necessary to evaluate improvement by testing in the real world for sustained periods of time. Plomp (See Reinier Plomp, *The Negative Effect of Amplitude Compression in Multichannel Hearing Aids in the Light of the Modulation-Transfer Function*, J. Acoust. Soc. Am. 83(6), June 1988, at 2322-2327) has suggested that multi-band compression would be detrimental to speech intelligibility because the reduction in dynamic range does not imply a reduction in the size of the just noticeable difference (JND) in amplitude discrimination. Plomp has further suggested that fast time constant compression would lead to reduced amplitude modulation over time, which in turn, would lead to reduced perception of this modulation. It has also been suggested that very fast time constants can create harmonic distortion at low frequencies. The argument was also put forward that fast time constant multi-band compression would reduce spectral contrasts over frequency, thus "whitening" the spectrum, thereby lessening the ability to distinguish vowels. Vilchur (See E. Vilchur, *Comments on the Negative Effect of Amplitude Compression in Multichannel Hearing Aids in the Light of the Modulation-Transfer Function*, J. Acoust. Soc. Am. 86(1), July 1989, at 425-428) responded to these points. Others have written on related topics. See e.g. L. D. Braida, et al.,

Review of Recent Research on Multiband Amplitude Compression for the Hearing Impaired, The Vanderbilt Hearing Report, Upper Darby, PA: Monographs in Contemporary Audiology, 1982, at 133-140; B. R. Glasberg, et al., *Auditory Filter Shapes in Subjects with Unilateral and Bilateral Cochlear Impairments*, J. Acoust. Soc. Am. 79, 1986, at 1020-1033; Brian Moore, *How Much Do We Gain by Gain Control in Hearing Aids?*, Acta Otolaryngol, 1990, 469 Suppl. at 250-256; Igor Nabelek, *Performance of Hearing-Impaired Listeners under Various Types of Amplitude Compression*, J. Acoust. Soc. Am. 74(3), Sept. 1983, at 776-791; and Walker et al., *The Effects of Multichannel Compression/Expansion Amplification on the Intelligibility of Nonsense Syllables in Noise*, J. Acoust. Soc. Am. 76(3), Sept. 1984, at 746-757.

Most agree that some form of limiting is required so that loud sounds are not too loud but soft sounds are audible. The debate is focused on full range vs. limiting compression, and on fast vs. slow time constants. Moore (previously cited) suggests that a two or three band compressor, while having sufficient frequency resolution to allow attenuation of low frequency noise and vowel sounds, and while permitting amplification of softer high frequency consonants, is still coarse enough in frequency, as opposed to a critical band compressor, such that spectral whitening will not occur.

Given two input signals of equal energy, one narrow band so that its frequency range is entirely within one critical band, and another wide band so that its frequency range spans several critical bands, the wide band signal will appear louder to the listener. This is due to a psychoacoustic phenomenon called loudness summation. This has implications for compressor design. If the compressor has a few wide bands (e.g. 2), and if the compressor is adjusted such that wide band signals are well matched in loudness to normal loudness growth curves, then narrow band signals will appear too soft. Conversely if the compressor has many independent narrow bands (e.g. critical bands), and if the compressor is adjusted such that narrow band signals are well matched in loudness to normal loudness growth curves, then wide band signals will appear too loud. Hohman (See V. Hohman, *Narrow/Wide Band Compensation in Coupled Narrow Band Aid*, Reihe 17: Biotechnik, Nr. 93, 1993, at 1-99) has designed a compressor which addresses this problem. It measures not only power but bandwidth of input signals and adjusts gain accordingly. This is called a coupled narrow band compressor.

As illustrated by the above discussion, different signal processing strategies have been developed to address different and specific hearing aid problems. In an attempt to increase the versatility of hearing aids, adjustable hearing aids have been developed. With adjustable hearing aids (which typically employ analog signal processors), certain parameters can be adjusted by the user. By allowing the user to dynamically set the parameters, the adjustable hearing aid allows the user to set the hearing aid to best suit the user's listening environment. While an adjustable hearing aid does impart to the user a greater degree of versatility, this versatility has its limits. Ultimately, an adjustable hearing aid implements the same signal processing strategy, regardless of the parameters. If the particular strategy implemented by the hearing aid is not well-suited for

a particular situation, then no amount of parameter adjustment will cause the hearing aid to provide satisfactory results.

Summary of the Invention

The present invention is based, at least partially, on the observation that, given all of the available signal processing strategies and all of the possible listening environments that a user may find himself or herself in, there is no single signal processing strategy which provides optimal performance in all situations. In order to be optimal, a hearing aid needs to implement different signal processing strategies for different situations, with each strategy designed for a particular situation. The present invention provides just such a hearing aid.

In accordance with the present invention, there is provided a hearing aid comprising a an input transducer, an analog-to-digital converter, a plurality of digital signal processing means, a processing means selector manipulable by a user, a digital-to-analog converter, and an output transducer. Preferably, each of the digital signal processing means implements a particular processing strategy designed for a particular situation. The processing means selector allows the user to select which digital signal processing means to invoke so that the user may dynamically choose the best processing means, and hence, the best strategy for any particular listening environment. In a preferred embodiment, the plurality of digital signal processing means of the hearing aid of the present invention is implemented by way of a logic unit and a multi-program store for storing a plurality of instruction sequences. These instruction sequences, when executed by the logic unit, cause the logic unit to implement the functions of the various digital signal processing means. To change digital signal processing means, all that needs to be done is to have the logic unit execute a different set of instruction sequence. Hence, the present invention provides a hearing aid capable of: (1) implementing a number of different signal processing strategies; and (2) allowing a user to select which strategy is implemented, thereby allowing the user to choose the best strategy for any given situation. Overall, the present invention provides a functionally superior hearing aid.

Brief Description of the Drawings

Figure 1a is a block diagram representation of the hearing aid of the present invention.

Figure 1b is a block diagram of a preferred embodiment of the hearing aid of Figure 1a.

Figure 2 is a functional block diagram of one of the digital signal processing means 50 of Figure 1a.

Figure 3 is a functional diagram of an incremental FFT filter bank in accordance with the present invention.

Figure 4 is a functional diagram of a hybrid filter bank in accordance with the present invention.

Figure 5 is a plot of the superimposed magnitude frequency responses of comb filters 301, 302 and 304, 306 of Figure 3.

Figure 6 is a plot of the composite frequency response seen at the output of adder 306 of Figure 3.

Figure 7 is a plot of the superimposed magnitude frequency responses of comb filters 301, 302 and 304, 307 of Figure 3.

Figure 8 is a plot of the composite frequency response seen at the output of adder 307 of Figure 3.

Figure 9 is a plot of the superimposed magnitude frequency responses of comb filters 301, 303 and 305, 308 of Figure 3.

Figure 10 is a plot of the composite frequency response seen at the output of adder 308 of Figure 3.

Figure 11 is a plot of the three superimposed frequency responses shown in Figures 5, 7, and 9.

Figure 12 is a composite frequency response of port out2 shown in Figure 3.

Figure 13 is a plot of the magnitude frequency response of a complex filter tuned to FSAMP/4 superimposed on the frequency response seen at the output of adder 406 shown in Figure 4.

Figure 14 is a plot of the composite frequency response seen at the output of the complex one-pole 441 shown in Figure 4.

Figure 15 is a plot of the group delay of a complex one-pole resonator with a .9 coefficient in a hybrid filter bank equivalent to a 256 point FFT.

Figure 16 is a block diagram representation of a bandsplitter in accordance with the present invention.

Figure 17 is a block diagram representation of an allpass bandmerger in accordance with the present invention.

Figure 18 is a functional representation of an arithmetic logic unit in accordance with the present invention.

Figure 19 is a functional representation of a first embodiment of the hybrid bandsplitter filter bank analyzer of the present invention, wherein a midband notch/bandpass filter is utilized.

Figure 20 is a functional representation of a second embodiment of the hybrid bandsplitter filter bank analyzer of the present invention, wherein a midband notch/bandpass filter is utilized, and wherein a real bandsplit is performed before converting to complex.

Figure 21 is a functional representation of a third embodiment of the hybrid bandsplitter filter bank analyzer of the present invention, wherein no midband notch/bandpass filter is utilized, and wherein a real bandsplit is performed before converting to complex.

Figure 22 is a functional representation of one of the one-band compressors of a multi-band compressor in accordance with the present invention.

Figure 23 is a more detailed functional diagram of the log smoother 2203 shown in Figure 22.

Figure 24 is a functional representation of a hybrid bandsplitter filter bank synthesizer/combiner corresponding to the analyzer shown in Figure 19.

Figure 25 is a functional representation of a hybrid bandsplitter filter bank synthesizer/combiner corresponding to the analyzer shown in Figure 20.

Figure 26 is a functional representation of a hybrid bandsplitter filter bank synthesizer/combiner corresponding to the analyzer shown in Figure 21.

Figures 27-32 are spectra patterns of various signals showing the results of decimation.

Figure 33 is a functional representation of a Hilbert transformer in accordance with the present invention.

Figure 34 is a functional representation of a system which results when the bandsplitter shown in Figure 16 is provided with a complex input and is partitioned into its real and imaginary parts.

Detailed Description of the Preferred Embodiments

With reference to Figure 1a, there is shown a block diagram representation of the hearing aid of the present invention, the hearing aid 40 preferably comprising an input transducer 42 (preferably taking the form of a microphone), an analog-to-digital converter 44, a selector switch 46, a plurality of digital signal processing means 50, each selectable by selector switch 46, a digital-to-analog converter 52, and an output transducer 54 (preferably taking the form of a speaker). The selector switch 46 is preferably manipulable by a user to allow the user to dynamically select which of the digital signal processing means 50 to invoke in which listening environment. Preferably, each of the digital signal processing means 50 is specifically and optimally designed to deal with a particular listening environment. For example, one of the digital signal processing means 50 may be designed to compensate for noisy environments, while another may be tailored for quiet environments. In dealing with these environments, each of the processing means 50 may implement such functions as compression, noise compression, feedback cancellation, etc.

The hearing aid of Figure 1a operates by receiving audio signals, via the microphone 42, from a particular listening environment, and converting these audio signals into a set of analog electrical signals. These electrical signals are then converted by the analog-to-digital converter 44 into an input digital signal or stream. The input digital stream is then fed to one of the digital signal processing means 50 selected by the selector switch 46, where the input digital stream is processed to derive an output digital signal or stream. This output digital stream is thereafter processed by the digital-to-analog converter 52 to derive a set of output analog electrical signals. Once derived, the analog electrical signals are used to drive the speaker 54 to cause the speaker to produce a set of audio signals which can be heard by the hearing aid user.

A significant advantage provided by the hearing aid of the present invention is that it is capable of implementing a plurality of different rehabilitation strategies. This is in sharp contrast to the adjustable hearing aids of the prior art which are capable of implementing only a single

strategy with adjustable parameters. Because the hearing aid of the present invention can implement more than one strategy (i.e. can change from one strategy to another), it is better able to adapt and to provide optimal results in a variety of different listening environments. As used herein, the expression "changing strategies" means generally switching from one digital signal processing means 50 to another. In actuality, what is often changed in going from one processing means to another is the number of bandpass signals into which the input digital signal is divided, and the bandwidths of these bandpass signals. Thus, by changing strategies, the hearing aid of the present invention is in effect changing the specifications of the filter banks of the digital signal processing means 50. This will become more clear as the invention is described in greater detail.

Referring now to Figure 1b, there is shown a preferred embodiment of the hearing aid of the present invention. Figure 1b shows the main components of the hearing aid 40. This architecture is suitable for implementing a number of dynamic range compression strategies as well as other hearing aid rehabilitation strategies. Sound is input through an input transducer (101) and converted to a digital input stream by the Analog to Digital Converter (102). Calculations are performed on data from the input stream as well as data stored in X Data Ram (105) and Y Data Ram (106). These calculation are carried out by the Arithmetic Logic Unit (108) with the Input Mux (107) selecting which sources of data will be processed. The results of the calculations are fed back to the X or Y Data Rams (105, 106) or to the Digital to Analog Converter (104) which converts the signal to an analog output electrical signal suitable for driving the Output Transducer (103). Corresponding to each hearing aid rehabilitation strategy is a digital signal processing program which is stored in the Multi-program Store (111). Each program corresponds to a set of instructions which are interpreted by the program sequencer (110) and executed by ALU 108 to cause various actions to take place within the rest of the circuit. The Program Selection Switch (112) selects which rehabilitation strategy to activate. This switch is under control of the hearing aid wearer so that as he or she enters different listening environments, the appropriate strategy can be selected. At this point, it should be noted that each of the digital signal processing means 50 shown in Fig. 1a is implemented in the preferred embodiment as a digital signal processing program executed by the ALU 108. By switching between processing programs using program selection switch 112, the hearing aid wearer is in effect switching between the plurality of digital signal processing means 50.

A number of rehabilitation strategies are implemented as digital signal processing programs stored in the Multi-program Store. Figure 2 shows the basic elements common to all of the rehabilitation strategies. The sound signal is input to a Filter Bank Analyzer (201) which divides the signal input into multiple frequency bands. The multiple frequency band signals are input to the multi-band processor (202) which processes the individual frequency band signals to affect their dynamic ranges. In the preferred embodiment, processor (202) performs a compression function; however, it should be noted that processor (202) may perform any desired function. The processed frequency band signals are then input to the Filter Bank

Synthesizer/Combiner (203) which recombines the individual frequency band signals into a single output. Since these basic elements are implemented as digital signal processing programs, and the system provides for a plurality of programs that the user can switch between, it is possible to change filter structures by selecting different programs. For example, some algorithms, such as noise suppression algorithms, may require fine frequency resolution filter banks, whereas more simple compressors may require only two bands. These differences can be accommodated by switching digital signal processing programs. The following sections describe embodiments related to these three basic components. One of the motivations for the various filter bank embodiments is to achieve high frequency resolution, especially at low frequencies, without incurring large delay through the system. This issue will be discussed in conjunction with the various embodiments.

INCREMENTAL FFT FILTER BANK ANALYZER

Figure 3 describes the first embodiment of a filter bank analyzer. The sum of a signal and a delayed version of itself form a comb filter with N filter lobes evenly spaced across frequency from 0 to the sample frequency (FSAMP). The peaks of the lobes of the comb filter are centered at k*FSAMP/N, with k ranging from 0 to N-1. The difference of a signal and a delayed version of itself also forms a comb filter with N evenly spaced lobes but now the peaks of the lobes are centered at (k*FSAMP/N)+FSAMP/(N*2), so that the lobe centers are shifted in frequency by half a lobe width. The magnitude frequency response of the sum of these two comb filters is flat, that is the comb filters are complementary and together they define 2*N frequency lobes ranging from 0 to FSAMP. A more general form of the comb filter is:

$$S + delay(S, N) \cdot cnn \qquad (1)$$

where S is the input signal, *delay()* is a function which delays the signal by input parameter N, and cnn is a multiplier coefficient defined as :

$$cnn = e^{(j\,2\pi \cdot N \cdot CENTER\_FREQUENCY / FSAMP)} \qquad (2)$$

which shifts the peak of the first frequency lobe to CENTER_FREQUENCY.

In Figure 3 the numbers c40, c41, c20, c21, c10, c11 etc. represent multiplier coefficients as defined in (2). Thus, the output of adder 302 is a comb filtered signal with delay N defined by 301 as 4. In this case c40=1 and c41=-1 so the output of adder 303 is the complementary comb filter with lobe centers shifted by FSAMP/(4*2). The output of 302 is then fed into another pair of complementary comb filters whose outputs are the adders 306 and 307 and multipliers c20=1 and c21=-1. The second stage comb filters defined by 304, 306, 307 and multipliers c20, c21 have lobe widths which are twice the width of the first stage comb filter with output 302 since 304 has delay 2 which is one half delay 301. Since the lobe width of the second stage comb filter 304, 306 is twice the first stage comb filter 301, 302, and since they both have their first lobes centered at frequency 0 then the second stage comb filter will have a zero at the peak frequency of the first stage comb filter's second lobe and all subsequent even number lobes. Figure 5 shows the superimposed magnitude frequency responses of comb filters 301, 302 and 304, 306.

With these two comb filters in series the composite frequency response seen at the output of adder 306 is shown in Figure 6. In effect the second stage comb filter selects lobes 1, 3 of the first stage comb filter and suppresses lobes 2, 4. The second stage comb 304, 307 is the complement of comb 304, 306 since c21=-1 and its lobes are shifted in frequency by half the second stage lobe with, that is by one of the first stage comb widths. Therefore, second stage comb 304, 307 selects lobes 2, 4 of the first stage comb and suppresses lobes 1, 3. Figure 7 shows the superimposed magnitude frequency responses of comb filters 301, 302 and 304, 307 and Figure 8 shows the composite frequency response seen at the output of adder 307.

The frequency response at the output of adder 303 is the complement of that at the output of adder 302 and is shifted by one half the first stage lobe width compared to the output of 302. The outputs of adders 308 and 309 are again complementary comb filters with lobe widths twice the width of the first stage comb, but now to select the even and odd lobes respectively of the output of adder 303 they must by shifted by .5 and 1.5 of the first lobe widths so that they will line up correctly with the output of adder 303. This means that c22=j and c23=-j where j=sqrt(-1). Thus the outputs of 308 and 309 are complex signals as will be the general case with this network. Figure 9 shows the superimposed frequency responses of combs 301, 303 and 305, 308 and Figure 10 shows the composite frequency response seen at the output of adder 308. At the output of adders 306 through 309 there are 4 frequency responses each one selecting two of the original (N*2=8) lobes defined by the first stage comb filters. By continuing this process of doubling the lobe width and shifting lobe centers, a third stage of comb filters with delays of 1 is added providing eight outputs, each selecting one of the original eight lobes as defined by the first stages comb filters. This requires complex multipliers c10-c17 selected appropriately. Figure 11 shows the three superimposed frequency responses and Figure 12 shows the composite frequency response leading to out2 which is the third lobe of the original eight.

The outputs of the system of comb filters defined in Figure 3 are identical to those of an 8 point Complex Fourier Transform. In effect the system is the implementation of an Incremental Complex Fourier Transformer with rectangular window. It is incremental because for every new input sample a new set of output transform samples is generated. By adding additional comb filter stages in front of the first stage shown in Figure 3, and by expanding the comb filter tree appropriately, longer Fourier Transforms can be calculated. The number of frequency points of the Fourier Transform is 2*N where N is the delay of the first stage comb.

The group delay of the Incremental Complex Fourier Transformer is the sum of the series interconnection of the combs. Each comb filter has a group delay equal to ½ the delay length. So for an N*2 point Fourier Transform the total group delay is N/2 + N/4 ...+ ½ which is equal to N-.5. Thus, the 8 point FFT system has a delay of 4-.5 = 3.5 samples. A typical block based implementation of a Short Time Fourier Transform system with a 2 to 1 overlap of successive FFT frames has a total delay of 3*(N) where N*2 is the FFT window length. This delay is due to system buffering requirements. This is more than 3 times the length of the optimal Incremental